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Multi-channel Magnetocardiogardiography System Based on Low-Tc SQUIDs in an Unshielded Environment

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Abstract

Magnetocardiography (MCG) using superconducting quantum interference devices (SQUIDs) is a new medical diagnostic tool measuring biomagnetic signals that are generated by the electrical activity of the human heart. This technique is completely passive, contactless, and it has an advantage in the early diagnosis of heart diseases. We developed the first unshielded four-channel MCG system based on low-Tc DC SQUIDs in China. Instead of using a costly magnetically shielded room, the environmental noise suppression was realized by using second-order gradiometers and three-axis reference magnetometer. The measured magnetic field resolution of the system is better than 1 pT, and multi-cycle human heart signals can be recorded directly. Also, with the infrared positioning system, 48 points data collection can be realized by moving the non-magnetic bed nine times.

Keywords: Magnetocardiography (MCG), SQUID, gradiometer

1. Introduction

Magnetocardiography (MCG) signal for adult human heart is very weak, typically several tens picoteslas, comparing to the strong environment magnetic noise, around several tens microteslas [1]. The technique for reducing the environmental noise is needed to measure MCG signal with efficiency. The most straightforward and reliable way of reducing the external magnetic disturbances is to perform the measurements in a magnetically shielded room (MSR) [2-4]. But, for its high cost and large volume size, MSR will become a major hurdle for the popularization of MCG system in clinical application. Therefore,

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many researchers are trying to investigate the system can measure the MCG signal without an MSR (unshielded) or with a moderated shielded enclosure [5-8].

In many cases, even in a shielded enclosure, mechanical vibration of the Dewar and the nearby noise sources, such as the electronics, will disturb the stability of the measurement, so that the simple magnetometers are rendered practically useless. In this case, different types of gradiometric layout have been considered in MCG system working in unshielded or moderated shielded environment [9-10].

In this paper, 2nd-order axial hardware gradiometer was fabricated with niobium wire, in which two 1st-order gradiometers are connected together in opposition so that the detection coil is insensitive to both homogeneous fields and uniform field gradients [11]. This arrangement can suppress the environmental noise to 32 dB and the signal to noise ratio of MCG signal is around -46 dB. With three orthogonal reference magnetometers to measure three field components [12] of the environmental background, the noise can be further suppressed to 40 dB and 60 dB with constant and/or adaptive coefficient cancellation, respectively. Finally, the MCG signal with signal-to-noise ratio of +13 dB is successfully detected in unshielded laboratory environment with adaptive compensation gradiometer technique [13].

2. Gradiometer Technique in MCG System for Noise Cancellation

In an unshielded laboratory environment, to measure the cardiac signal, which is of several tens picoteslas, gradiometer technique is a widely used method to suppress the environmental interferences. The earth field is about 40 microteslas. Therefore, the ability of noise reduction in unshielded environment should be more than 100 dB to detect human heart magnetic field signal with high signal-to-noise ratio (SNR). In our system, we choose a 2nd-order hard gradiometer and reference 3-axis magnetometer module [12] to combine a software gradiometer.

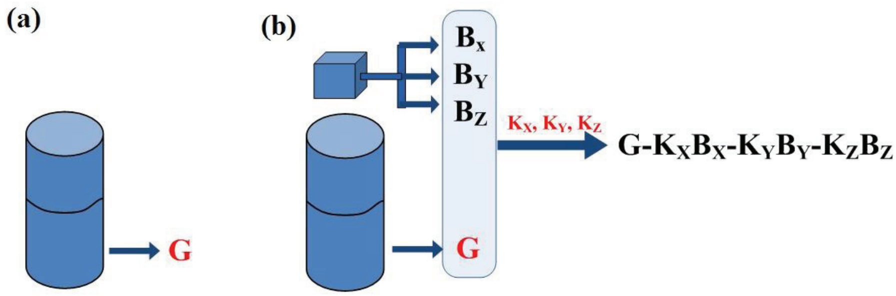


Fig.1. Different gradiometer configurations (a) 2nd-order hardware gradiometer; (b) 2nd-order hardware gradiometer and three-axis reference magnetometer to combine an electronic gradiometer when K_x, K_y and K_z are constant, or to combine an software gradiometer when K_x, K_y and K_z are adaptive.

Fig.1 shows the different gradiometer configurations: (a) is a pure 2nd-order axial gradiometer; (b) is a gradiometer combining a pure 2nd-order axial gradiometer and a three-axis reference magnetometer. So the output of the above configuration will become $V_o=G-K_xB_x-K_yB_y-K_zB_z$, in which G is the output of a pure 2nd-order axial gradiometer, B_x, B_y and B_z is the field output of three-axis reference magnetometer and K_x, K_y and K_z are coefficients. When the coefficients are constant or adaptive, an electric gradiometer or a software gradiometer will be constructed, respectively.

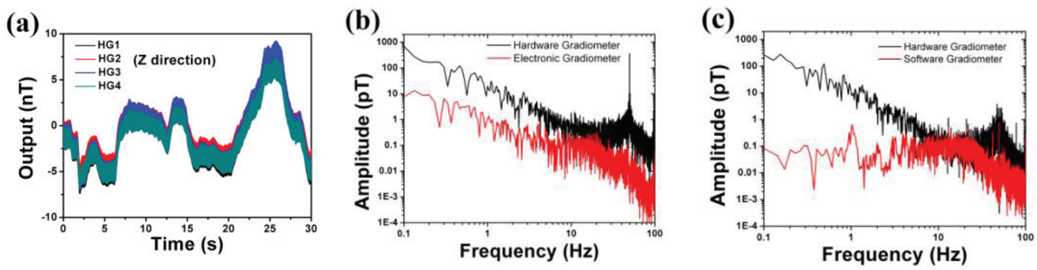


Fig.2. (a) Output of 2nd-order hardware gradiometer when measuring the environmental field and the comparison of noise suppression (b) between hardware gradiometer and electronic gradiometer; (c) between hardware gradiometer and software gradiometer.

Firstly, a 2nd-order axial gradiometer is fabricated with ϕ 0.1 mm niobium wire wounded of 1-2-1 configuration on a macor bar with diameter of 18 mm. The baseline of the axial hardware gradiometer is 50 mm. The inductance of the 2nd-order pickup coil is 607.7 nH. The inductance of the SQUID input coil connection with it is about 683 nH and $1/M=0.2\ \mu A/\Phi_0$.

Fig.2. is the z component output of four-channel axial hardware gradiometers in the unshielded environment without the reference compensation. From the curves we can see that the hardware gradiometer can work stably in unshielded environment with noise suppression of 32 dB. But the output is still too high for MCG signal detection.

If we use three-axis reference magnetometer to compensate the imbalance response of hardware gradiometer, Fig 2(b) and (c) are the output of electronic gradiometer and software gradiometer, whose coefficient is constant and adaptive respectively [13]. From the result, we can see that the FFT is much lower than that of hardware gradiometer after 0.1-95 Hz bandpass filter. In Fig.2 (b), the output level of electrical gradiometer is lower than that of hardware gradiometer with noise suppression about 40 dB. In Fig.2 (c), the software gradiometer is much better than hardware gradiometer, especially in 10 Hz and less, noise suppression can be up to 60 dB. The reason is that the coefficients in electrical gradiometer are adjusted by circuit components and the response of 3-axis reference magnetometers cannot trace the change of environment. The software gradiometer has more advantage in this point using the smoothing RLS algorithm to adaptively change the coefficients with environment. Also, the level at 50 Hz is well suppressed with notch filter.

After electronic gradiometer and software gradiometer, the noise reduction will be 72 dB and 92 dB, respectively. Fig. 3 is the comparison of the output of four channels with MCG signal after the above two gradiometers. From fig.3 (a), we can also see that there is low frequency drift in the output of electrical gradiometer, while the base level is very flat of software gradiometer output. From the enlarged graph in fig.3 (b), the R and T peak can be seen obviously. There are still some high frequency noise and separate interference such as 21 Hz, 42 Hz. Also the suppression ability is limit for the software gradiometer.

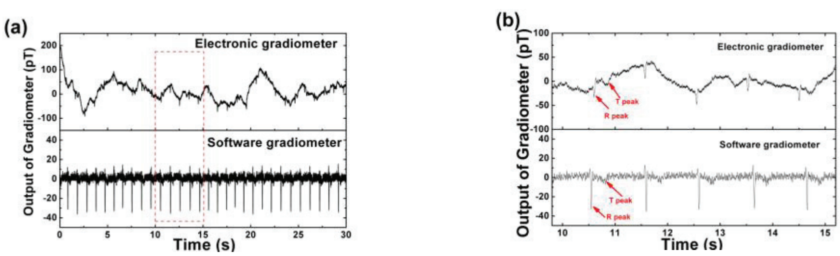


Fig.3. MCG signal detected in unshielded environment with electronic gradiometer (top) and software gradiometer (bottom).

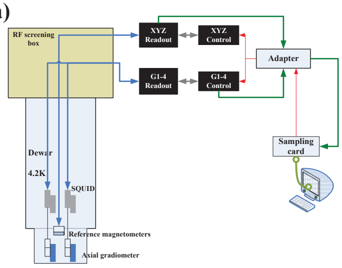
To evaluate the signal integrity, a simple yet effective method was developed by using the electrocardiographic (ECG) signal to drive a small coil to generate a simulated MCG signal. Three key performance indexes were proposed, which are correlation in time domain, relative heights of different peaks and correlation in frequency domain. The signal-to-noise ratio of the extracted multi-cycle and averaged MCG signal were measured to be 5 and 80 respectively, and the correlation coefficient between the output MCG signal and the simulated ECG signal was up to 0.999 [14].

3. MCG System and Evaluation

3.1 System setup

The schematic diagram of our MCG equipment in unshielded laboratory is depicted in fig. 4(a), which consists of nonmagnetic movable patient table, a liquid helium fiberglass dewar, in which 7 channel low-Tc DC SQUID sensors (4 signals and 3 references) are placed at the bottom of it and immersed in liquid helium. The SQUID is connected to the circuit of flux locked loop (FLL) at room temperature via a matching circuit and a cryogenic cable. We hang the dewar on a framework and adjust the vertical position of the dewar from above as close as possible to the patient's chest. Generally, the vertical distance between sensor and chest is around 15 cm. In addition, an electrocardiogram (ECG) measurement can be made simultaneously with MCG measurement and the ECG data can be taken by the computer. Radio frequency should be well shielded for the total MCG system, such as super-insulating layer outside liquid helium cryostat, the PFL boxes and transmitting wires, to reject the high frequency interference in the environment. The photo of the MCG system is shown in fig. 4(b).

(a)



(b)

Fig.4. (a) The schematic configuration of the MCG system based on low-Tc DC SQUID system; (b) the photo of the above system.

The data can be acquired with PC controlled system with high frequency sampling 20 kH, and doing 130 Hz low pass filter and resample (1 kHz). MCG signals of high SNR will be obtained by pre-precess using a 100Hz low pass filter and 50 Hz notch filter to cancel the high frequency and power line. Then the noise from the gradiometer imbalance will be cancelled with an optimized. Finally，single period MCG signal was achieved by carrying out multi-cycle average to suppress the residual high frequency components by using the ECG R peak as a time reference.

3.2 Evaluation of MCG system

In multi-channel MCG system, the crosstalk is one of the most important issues to be considered. The crosstalk includes two parts: crosstalk from readout circuits and that from magnetic field [15]. The crosstalk between circuits comes from the difference of FLL modulation frequency. It can be solved by the synchronous modulation for the four signal channels. The relative output difference between only one channel work and that four channels work at same time is around 0.106%. The crosstalk of magnetic field mainly comes from the coupling between pickup coils belonging to the SQUID flux loop. The measured result is around 0.067%. But for more channel system, it will be much higher and can be reduced by current feedback mode scheme.

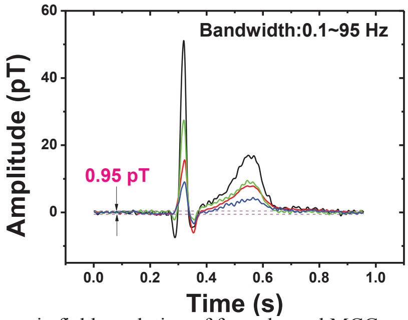


Fig.5. The magnetic field resolution of four-channel MCG system in unshielded environment.

The calibration of the multi-channel MCG system is to obtain the magnetic field to voltage transfer function ∂B/∂V. We use a large coil in two steps. Firstly, a large circular source coil is put at the central axis of the four signal channel under the dewar, near the pickup coil of the axial gradiometer. By comparing the output voltage of four signal channel, we can get the ratio of the 4 channels 1.000:1.019:1.013:1.024. Second step, we move the large coil to the position right under pickup coil of channel 1. By measuring the output of the SQUID and calculating the magnetic field at the same position, the resolute transfer coefficient will be 0.669 mV/pT. All the transfer coefficients are 0.669 mV/pT, 0.682 mV/pT, 0.677 mV/pT and 0.685 mV/pT for the four signal channels, respectively.

Fig. 5 is the magnetic field resolution of four-channel MCG system in unshielded environment on the basis of the data before P peak. The SNR of averaged one cycle MCG signal is up to 80. The bandwidth of the system is around 0.1-95 Hz as shown in the figure 5.

3.3 MCG imaging using multi-point data

During the measurements, the magnetic signal detected by the SQUID sensor from patient heart beating was linearized in the flux locked operation mode of the device. Then the signal was processed by a band-pass filter and 50 Hz notch filter digitized with a 14-bit A/D converter successively. The digital data were acquired at a rate of 2 kHz and further treated in the computer. By sequentially adjusting the patient table with 3.5 cm pitch in X and 4cm in Y directions, the field signals on a regular grid 8 × 6 points over the chest area of 24.5×20 cm2 were recorded in real time trace point by point with a typical dwell time over 30 cardiocycles. Based on the four-channel MCG system, a 36 grid measurements were performed covering a thorax area of 24.5 × 20 cm^2. Waveforms chart, magnetic field maps (MFM), pseudo current density maps (PCDM) etc, were obtained as shown in fig. 6 (a), the interface of post-processing. Fig. 6 (b) is the enlarged MCG signals at different 48 points.

(a) (b)

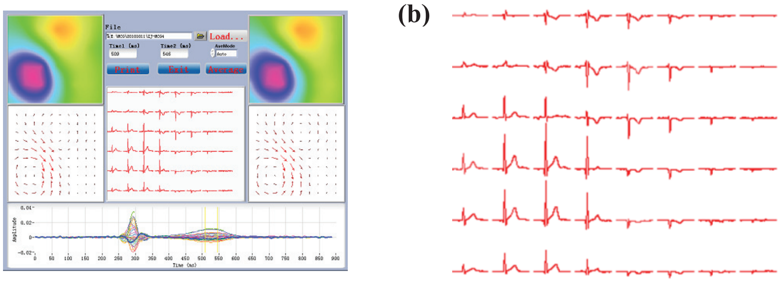


Fig. 6. (a)The post-processing interface for multi-channel MCG system; (b) The enlarged averaged MCG signals at 48 points with the vertical and horizontal space of 3.5 cm and 4 cm in horizontal and vertical direction, respectively (area: 24.5 cm × 20 cm) .

4. Conclusion

In this paper, a multi-channel MCG system based on low-Tc DC SQUID was setup in the unshielded laboratory in China for the first time. A 2^{nd}-order hardware gradiometer combines with a three-axis reference module to a software gradiometer, which has the noise suppression ability around 100 dB. The field sensitivity is around 1 pT and the bandwith is 0.1-100Hz. Using the 2-D non-magnetic moving bed and the orientating system, 48 MCG data array on the area of 20 cm × 24.5 cm have been obtained. The cardio-imaging for the above MCG data is carried out and the magnetic field maps (MFM), Pseudo Current Density Maps (PCDM) and MCG waveforms have been displayed. From above mapping, useful clinical parameters can be obtained.

For the MCG system working in unshielded environment, the stability and the noise suppression ability is of great importance. In the near future, we will further improve the stability and the SNR of MCG signal. Also, developing a MCG system with more channels and carrying out clinical application are the tough task in the near future.

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